Total knee replacement musculoskeletal model using a novel simulation method for non-conforming joints

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TOTAL KNEE REPLACEMENT MUSCULOSKELETAL MODEL USING A NOVEL SIMULATION METHOD FOR NON-CONFORMING JOINTS

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SUMMARY

We computed the medial and lateral knee compressive forces for a total knee replacement patient during gait based on the grand challenge data set. The model was based on a scaled version of the Twente Lower Extremity Model (TLEM) with the knee altered to include the prosthesis geometry (the femoral part, the tibial insert and the patella button) as well as ligaments. Contact conditions between the prosthesis parts were established using contact forces. The motion in the internal degrees-of-freedom (DOF) of the knee, the muscle and reaction forces were computed during stance of gait using a novel method called force-dependent kinematics (FDK) implemented in version 5.0 of the AnyBody Modeling System (AnyBody Technology A/S, Denmark).

INTRODUCTION

Design of prosthetic devices requires knowledge of the loading conditions that the prosthesis is exposed to throughout its lifetime. Due to the impossibility of directly measuring all forces affecting an implant, especially in the early design phase, estimation of the forces using a musculoskeletal model is the only viable solution. However, most musculoskeletal models presume idealized joints, such as a revolute joint for the knee and a spherical joint for the hip [1,2]. This is convenient in inverse dynamic analysis, where the kinematics of the model is resolved first, i.e. without knowledge of the forces that created the motion. However, several anatomical and prosthetic joints are non-conforming to such an extent that the forces significantly influence the detailed joint kinematics and the joint's internal force equilibrium. This is the case for joints such as spinal disks, knees and many shoulders. For instance, in the knee the internal motions are governed by a complex interaction between the muscle actions, cartilage contact mechanics, ligament forces and soft tissue deformations. Capturing all these effects in a realistic model using only kinematic constraints is very difficult, if not impossible. In this study, we take a radically different approach to joint modelling and apply the novel FDK method to compute the medial and lateral compressive forces, muscle and reaction forces while taking into account the joint geometry as well as the elasticity of the ligaments.

METHODS

A lower extremity musculoskeletal model was constructed in the AnyBody Modeling System v. 5.0 using the TLEM model [1] and the data set made publically available under the Grand Challenge [3] to predict the compressive forces in the medial and lateral compartment of the knee during normal as well as trunk sway walking. This data set includes measured knee medial and lateral compressive forces from an instrumented prosthesis, Computerized Tomography (CT), measured Electromyography (EMG) and trajectories of motion capture markers among others for one male test subject (Age: 83 years; Height: 166 cm; Weight: 64,6 kg).

The TLEM model is a detailed cadaver-based model of the lower extremity, which includes approximately 160 muscle units. It includes the thigh, patella, shank and foot segments, where the hip is a spherical joint and the knee, ankle, subtalar and patellofemoral joints are revolute joints.

A novel modelling approach, named force-dependent kinematics (FDK), capable of simultaneously computing the muscle and reaction forces using muscle recruitment typical for inverse [2] and the resulting motion in user-defined directions, was used. These motion directions must cover the degrees-of-freedom (DOF) in the knee that are not significantly accurately defined by the motion capture data, i.e. the small motion of the internal DOFs. The method is based on an assumption of quasi-static force equilibrium between all the acting forces in the model in these directions of motion; in other words, the method neglects the dynamics of these small motions and thereby assumes the stiffness in these directions to be high and the motion to be so small that it does not significantly influence the gross motion of the system, while allowing for displacements balanced by passive-elastic forces.

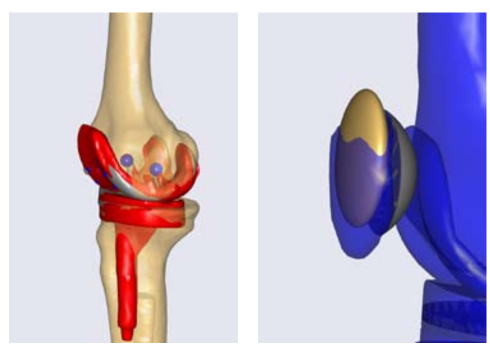

Figure 1: Alignment of the prosthesis. Left: The femoral part (gray) and tibial insert (gray) was aligned to the CT scan of the patient (red) and the TLEM model using the centers of the epicondyles and an estimation of the flexion/extension angle. Right: Alignment of the patella button (gray) relative to patella in the model (yellow) using the CT scan (blue).

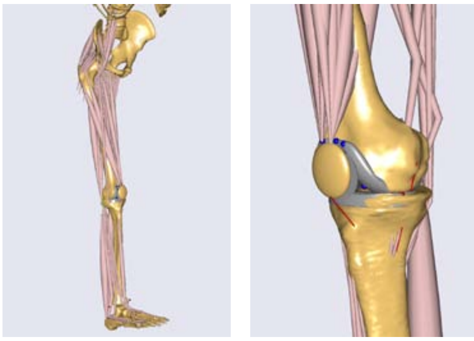

Figure 2: The lower extremity model with the included prosthesis and ligaments.

First, the standard TLEM model was scaled using a length-mass-fat scaling law [4] and the local motion capture marker coordinates computed using the method of Andersen et al. [5] over a gait trial. From this model, the scaled segment lengths as well as the generalized joint coordinates were saved for later use. Hereafter, the revolute joint knee model was removed and replaced by a detailed model, including contact mechanics, based on the geometry of the instrumented prosthesis as well as ligaments. The ligaments were modeled as linearly elastic with a given slack length. According to the grand challenge data description, both the ACL and PCL were removed during surgery. Therefore, only the medial and lateral collateral ligaments were included. The patella tendon was presumed infinitely stiff.

The implant geometry was aligned to the geometry of the scaled TLEM bone geometry using CT scans (see Figure 1) and the model is illustrated in Figure 2.

Finally, the internal motion, muscle and reaction forces of the tibiofemoral joint in all three translational directions as well as the internal/external rotation and abduction/adduction rotations and the patellofemoral joint in the anterior/posterior direction were computed using the FDK solver. The remaining DOFs were driven using the saved generalized joint coordinates.

RESULTS AND DISCUSSION

The computed and measured medial and lateral contact forces over the stance phase are plotted in Figure 3. The medial compressive force shows two peaks both in the measurement and in the model. The first peak shows an estimation error of 30 N. The second medial peak is significantly overestimated by the model with a peak error of 614 N. The computed lateral force also shows a trend similar to the measurement, but generally overestimates the force, especially the peak force around 80 % stance, where the error is about 663 N. Overall, the computed medial and lateral compressive force shows a root-mean-squares (RMS) errors of 241 N and 309 N, respectively.

There are several possible explanations of the errors in the computed compressive forces. One possible explanation is that the model muscle and bone morphologies were not altered to accurately reflect the patient due to the simple scaling method

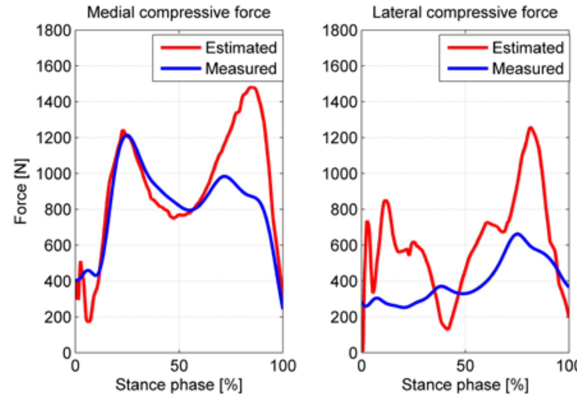

Figure 3: The computed and measured medial and lateral compressive knee forces.

applied. This could significantly affect the muscle moment arms, which results in inaccuracies in the joint forces. Experiments with the model show that a likely explanation for the estimation error is related to the moment arms of the Gastrocnemius and Rectus femoris, which are both biarticular muscles. Another explanation could be errors in the modeling of the ligament stiffnesses both in terms of the applied linear model and its model parameters. Furthermore, the errors could also originate from inaccuracies in the motion capture data, which is affected by soft tissue artifacts, among others. A significant problem with the motion capture data is that the foot markers were placed on a shoe. Especially, the toe marker is problematic because it was placed on the very tip, where the shoe flexes, which is not captured by the rigid foot model.

CONCLUSIONS

In this study, we applied a novel simulation method capable of simultaneously computing muscle and reaction forces as well as the motion in user-defined DOF. This approach enables detailed modeling of joints with passive-elastic forces.

Preliminary computed medial and lateral compressive forces in a model of TKR were presented. Throughout the first 50 % and the last 10 % of the stance phase of gait, the computed medial compressive force corresponds well with measurements, whereas the last 40 % of the stance phase, the force is over predicted. On the lateral side, the compressive force is generally over predicted by the model. Further studies are required to improve the model predictions.

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